

Fast-Fourier transform analysis of signal-averaged electrocardiograms for identification of patients prone to sustained ventricular tachycardia*

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ABSTRACT Electrocardiograms obtained from patients during arrhythmia-free intervals do not identify those prone to sustained ventricular tachycardia (VT) despite the occult delayed activation that is presumably present. To determine whether frequency-domain analysis facilitates detection of this hallmark of predisposition to VT, fast-Fourier transform analysis (FFTA) procedures were developed and tested with a computer-generated mathematical model. The FFTA approach developed allows inherent limitations of high-gain amplification and a priori filtering used commonly for time-domain analysis to be avoided. After demonstrating that FFTA detected low-amplitude oscillatory waveforms in signal-averaged recordings in the frequency domain, the procedure was applied to signal-averaged X, Y, and Z lead recordings from the following three groups of patients: group I, patients with prior myocardial infarction and episodic sustained VT ($n = 16$); group II, patients with prior myocardial infarction without overt sustained VT ($n = 35$); and group III, normal control subjects ($n = 10$). Results of FFTA demonstrated significant ($p < .0001$) differences in the decibel drop at 40 Hz and the area under the curve from the fundamental frequency to the frequency at which the spectral amplitude was decreased by 60 dB for both the terminal 40 msec of the QRS and ST segment in patients in group I compared with those in groups II and III, in whom results were similar. Results were independent of QRS duration ($r = .2$), left ventricular ejection fraction ($r = .19$), and complexity of spontaneous ventricular ectopy. Thus, patients known to manifest sustained VT also exhibited relatively greater high-frequency content in arrhythmia-free intervals in the terminal QRS and ST segment than those without VT (88%, 15%, and 0% in groups I through III, respectively). FFTA offers promise for the noninvasive detection of patients at risk for the development of sustained VT.

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ACCURATE NONINVASIVE DETECTION of patients at risk for development of sustained ventricular tachycardia (VT) is not yet possible. Recently, high-gain amplification and signal-processing techniques in the time domain have detected low-amplitude, high-frequency potentials in the terminal QRS complex and ST segment of signal-averaged electrocardiograms (ECGs) obtained during arrhythmia-free intervals from patients and experimental animals with sustained VT.¹⁻⁸ Results of studies in which endocardial and epicardial mapping were used suggest such potentials

reflect either delayed or disorganized ventricular activation.⁹⁻¹¹

Fast-Fourier transform analysis (FFTA) is a powerful analytic method for signal processing in the frequency domain that allows some of the inherent limitations of high-gain amplification and signal filtering required for analysis in the time domain to be avoided.¹² Any periodic signal, such as the QRS complex, may be represented by the mathematical summation of a series of sine waves of differing frequencies and amplitudes. The sinusoidal component with the lowest frequency is called the fundamental and has a repetition rate equal to the repetition rate of the periodic signal under evaluation. All higher sinusoidal components, or harmonics, have frequencies that are integer multiples of the fundamental frequency. FFTA is a computer-based mathematical algorithm whereby the amplitudes of the various harmonic components that comprise a complex periodic waveform are determined. The Fourier transform is unique since for each

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time-domain signal there is one and only one frequency domain presentation and vice versa.

To determine whether FFTA would facilitate objective identification and characterization of abnormal low-amplitude potentials in the surface ECG, procedures for FFTA of signal-averaged ECGs were developed, rigorously tested with both computer-generated mathematical functions and analog test voltages, and subsequently applied to patients. Results of FFTA of signal-averaged ECGs in patients with prior myocardial infarction and at least one documented episode of sustained VT were compared with those from patients with prior myocardial infarction without sustained VT and with those from normal subjects.

Methods

Computer-generated mathematical model. To test the hypothesis that FFTA may facilitate identification and characterization of high-frequency, low-amplitude potentials in the signal-averaged ECG, we first implemented a computer-based mathematical model and then subjected it to rigorous testing before constructing the signal-processing unit for patient use. To mathematically simulate a small oscillatory waveform superimposed on the trailing edge of a QRS complex, two complete cycles of a 40 Hz sine wave of variable voltage and onset were superimposed on a ramp function with a negative voltage vs time slope starting at a fixed voltage of 1 mV and reaching 0 mV at 50 msec. The use of FFTA for a discrete sample of a periodic waveform requires the assumption that the signal is a repetitive function and the initial and final sample points are at a potential of zero. However, if the initial and final sample points are not zero, a sharp-edge discontinuity will be introduced between the end of one cycle and the beginning of the next that will artifactually add both high and low frequencies to the original signal. To eliminate this source of harmonic error, time-domain samples are multiplied by a window function that

smooths the initial and final sampling points to zero at the boundaries, allowing periodic extension of the finite signal.¹³ In the computer-generated mathematical model both the ramp function and 40 Hz sine wave were premultiplied by a first-order Hanning window function to avoid edge discontinuity (figure 1). A first-order function was chosen to provide mathematical simplicity. Classic Fourier series analysis was performed on both functions, the spectra generated being verified by time-domain reconstruction of the original waveform (figure 2). The power spectrum of the sum of the ramp and sine burst function is simply the sum of individual components (sine and cosine terms added separately) of their respective spectra since the Fourier series transform is linear. To test the sensitivity of the FFTA the amplitude of the sine burst function was then decreased from a peak-to-peak level of 1 mV to 1 μ V and its effect was visualized in the frequency and time domains. The composite signal of the time domain was also displayed with selected levels of gain and prefiltering in an attempt to visualize the small oscillatory component that was added to the large simulated QRS trailing edge (figures 2 and 3).

Patients studied. Sixty-one patients were grouped according to their clinical characteristics. None was receiving antiarrhythmic medication at the time of study. Group I comprised 16 patients, each of whom had prior myocardial infarction and at least one documented episode of sustained VT or cardiac arrest that was not associated with a new infarct. All of the 13 patients studied in the clinical electrophysiology laboratory had inducible sustained VT or ventricular fibrillation similar to that occurring clinically. Group II consisted of 35 patients, each of whom had prior myocardial infarction but without a history or documented episode of sustained VT (> 30 sec in duration), syncope, or cardiac arrest, who were admitted to the Barnes Hospital Telemetry Unit. All patients in this group were monitored for at least 7 days. Seventeen patients exhibited absent or simple ventricular ectopy (modified Lown class 0 to 1), 18 had complex ventricular ectopy (class 2 to 5), and nine of these had nonsustained VT. Group III included 10 healthy male control subjects who were 24 to 40 years of age and in whom there was no clinical evidence of organic heart disease or arrhythmia.

Pertinent clinical features of patients in groups I and II are summarized in table 1. There were no significant differences

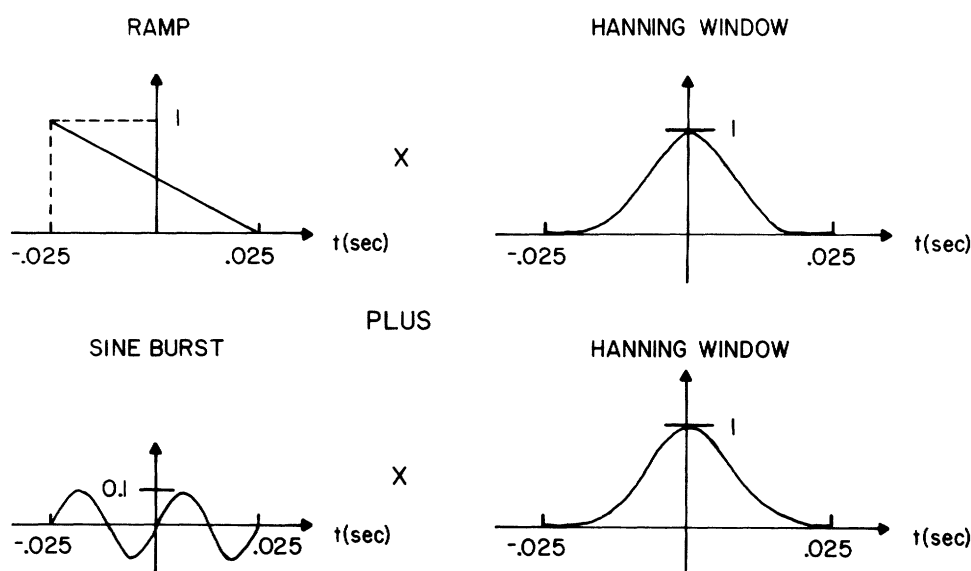


FIGURE 1. Time-domain representation of the ramp and 40 Hz sine burst functions used to simulate a small oscillatory waveform superimposed on the trailing edge of the QRS complex. To exclude edge discontinuities both functions are multiplied by a first-order Hanning window function before frequency-domain analysis.

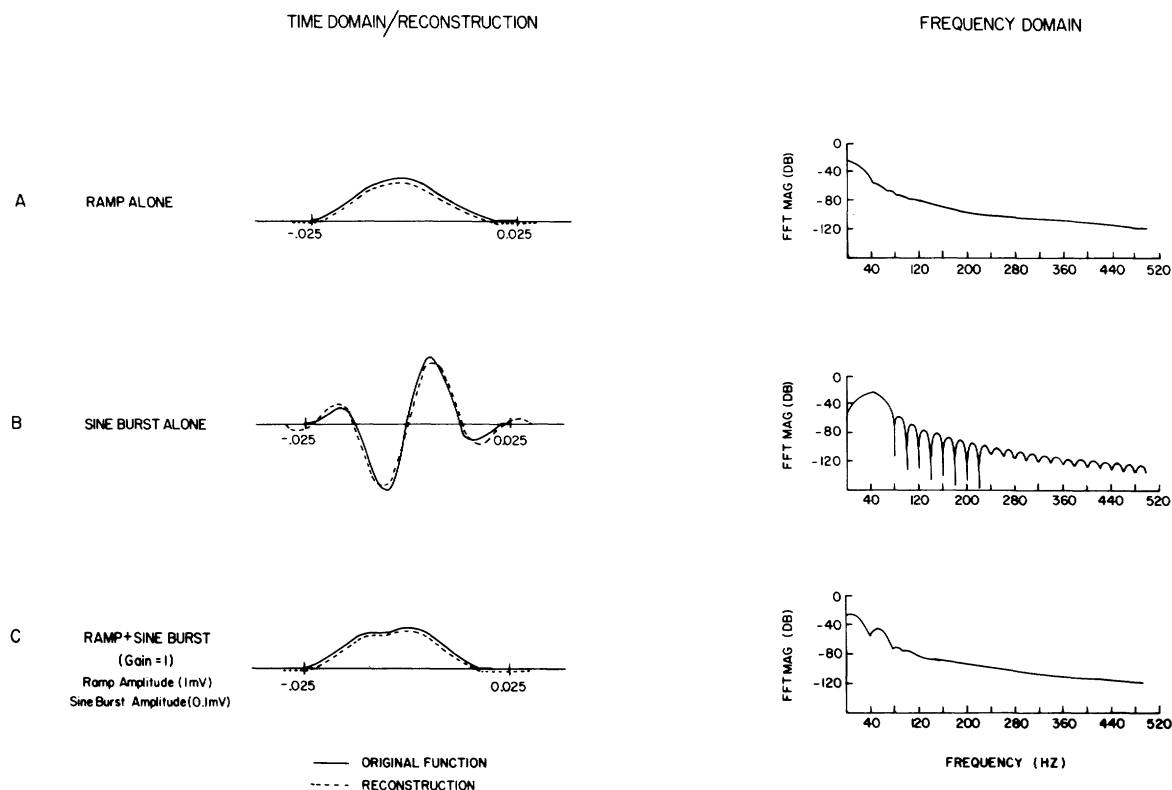


FIGURE 2. Comparison of results of time- and frequency-domain analyses. *Right.* The spectral analysis (after Hanning-window multiplication) of the ramp function (A), 40 Hz sine burst function (B), and the sum of the ramp and sine burst functions (C) are shown. *Left.* The original time-domain waveforms (solid lines) of these functions are shown along with the time-domain reconstruction (broken lines) from the Fourier series. C. The amplitude (0 to peak) of the sine burst is 0.1 mV, and the peak amplitude of the ramp function is 1 mV. The time-domain perturbation that reflects the sine burst contribution is minimal without prefiltering. However, the contribution from the 40 Hz sine wave is easily detectable in the frequency domain.

between these two groups with regard to age, infarct location, presence of left ventricular aneurysms, or QRS duration. Left ventricular ejection fraction was significantly less in group I patients compared with in group II patients ($34 \pm 16\%$ vs $45 \pm 15\%$; $p < .02$).

Surface ECG recording. Standard Frank bipolar X, Y, and Z ECG lead signals were amplified approximately 1000-fold (Hewlett-Packard 1057A) at 0.05 Hz to 470 Hz (noise level $\pm 15 \mu\text{V}$). The amplitude of the analog ECG signal was optimized to encompass the maximum $\pm 2.5 \text{ V}$ input range of the A/D converter. Normalized signals were then A/D converted at 1 KHz (12-bit accuracy), with a lowest resolution of $1 \mu\text{V}$. The digitized signals were processed with a portable cart with a DEC VT 103 LSI 11/23 microcomputer system with 64 K bytes of memory, two serial communication ports, a dual floppy disc system, and a Selanar raster graphics board with joystick control. All recordings were made at the patient's bedside at an interval of approximately 10 min.

Signal averaging. The X, Y, and Z ECG signals were averaged after passage through a template recognition program generated from a 3 sec display of normal sinus rhythm. The lead having the largest R wave amplitude (usually the X lead) was selected, the R-R interval and fiducial point (peak of the R wave) set with an adjustable cursor, and the QRS amplitude calculated for all three leads. Reference jitter was $\pm 1 \text{ msec}$. All subsequent beats were tested against the template with a cross-correlating technique and accepted for averaging only if: (1) the R-R interval was within $\pm 20\%$ of the previously set R-R interval, (2) the QRS amplitude had not changed in two of the leads, and (3) the fiducial point ± 20 sample values on either side of

the fiducial point had a correlation coefficient $\geq 98\%$ in comparison with the template beat. The beat immediately after a rejected beat was rejected also. Data from 100 beats were averaged and stored on a floppy disc for further processing. A 100 beat average reduced the inherent noise level from 15 to $1.5 \mu\text{V}$.

FFTA. FFTA was performed on the entire QRS complex, the terminal 40 msec of the QRS complex, the ST segment, and the T wave of each signal-averaged X, Y, and Z lead. Each component was identified with the use of the computer graphics cursor and standard ECG criteria (figure 4). For each region of interest, a 512-point fast-Fourier transform was calculated with the use of a four-term Blackman-Harris window (figure 5) with a 92 dB sidelobe level and sidelobe falloff of 6 dB per octave.¹³ A window function reduces spectral leakage due to edge discontinuities associated with analysis of a discrete subset (i.e., terminal 40 msec of the QRS complex) of a complex periodic waveform. Spectral leakage is most marked for the detection of low-amplitude signals in the presence of nearby large-amplitude signals. The ability of a window function to diminish spectral leakage is directly related to its sidelobe level. With the four-term Blackman-Harris window, the contribution of artifactual frequencies due to spectral leakage are apparent only at decibel drops greater than 92 dB. After multiplication by the window function, the selected sample values or regions of interest were placed at the beginning of the 512-point input array and the remaining values set to zero. This step permitted maintenance of the same frequency scale in the output data but allowed a varying number of input values to be analyzed with the same software. Since data were obtained at 1 msec intervals, samples of up to 512 msec in length could be analyzed. After FFTA the data

EFFECT OF AMPLIFICATION AND PREFILTERING

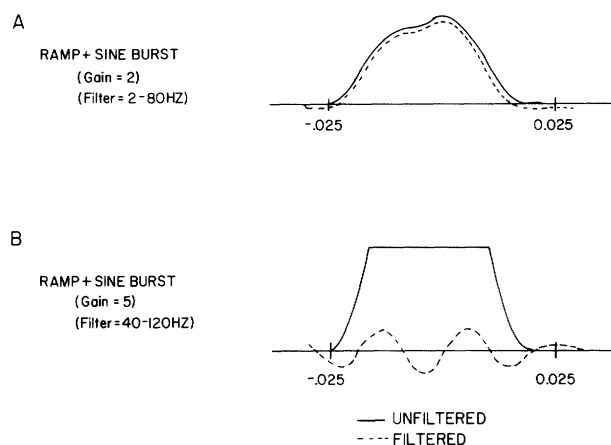


FIGURE 3. Effect of gain and prefiltering in the time domain. In *A* and *B*, the original time-domain waveform (solid line) and the idealized filtered waveform (broken line) are shown. *A*, The display gain is twice that in figure 2, *C*. When the filter used in the reconstruction is 2 to 80 Hz, the filtered waveform and original waveform correlate closely. However, the sine burst contribution is still only minimally apparent. *B*, The display gain is five times that in figure 2, *C*. The filter used in the reconstruction is 40 to 120 Hz. With high-gain amplification and prefiltering the sine burst can be adequately visualized in the time domain. However, selection of the high- and low-pass filters requires a priori knowledge of the frequency content of the signal of interest.

were plotted on a high-resolution plotter (Versatec, Inc.) and transferred to a disc for storage.

Validation of the system for use with patients. To evaluate the portable signal processing system before use in patients an analog test signal was constructed consisting of a 40 Hz sine wave of variable voltage phase locked to a triangular wave with a repetition rate of 1 Hz. The triangular waveform simulated a QRS complex that was recognizable by the signal-averaging algorithm, and thus acted as a point of time synchronization for

TABLE 1
Characteristics of patients

	Group I (16 patients)	Group II (35 patients)
Age (years)	59 ± 9.7	61 ± 10.2
LVEF (%)	34 ± 16 ^Δ	45.2 ± 15.2
QRS width (msec)	111 ± 27.3	88 ± 15.9
MI location		
Anterior	9 (56%)	22 (63%)
Inferior	7 (44%)	13 (37%)
Highest-grade VEA documented		
Simple	—	17
Complex	—	9
Nonsustained VT	—	9
Sustained VT	16	0

Data are expressed as mean ± SD.

MI = myocardial infarction; LVEF = left ventricular ejection fraction; VEA = ventricular ectopic activity.

^Δp = .02.

the sine wave signal. The analysis window was chosen to incorporate 50 msec of the sine wave, simulating a sine wave burst similar to that used in the computer-generated mathematical model described previously. To characterize the amplitude resolution of the system, the amplitude of the sine burst function was decreased from a peak-to-peak level of 1 mV to 1 μ V. The signal-averaging algorithm was evaluated by measuring the decibel drop in signal noise when the sine wave burst was averaged for 100 compared with one cycle.

Data analysis. For each region of interest (QRS complex, terminal QRS, ST segment, and T wave) spectral plots were generated for the ECG leads X, Y, and Z. The spectral plots were normalized independently of input amplitude by representing the fundamental frequency as 0 dB. From each spectral plot the computer defined the decibel drop at 40 Hz and the area under the curve from the fundamental frequency to the frequency at which the amplitude of the spectrum was 60 dB lower than peak (60 dB area; figure 4). The 40 Hz intercept was chosen because most of the energy of a normal QRS is less than 35 Hz^{14, 15} and because work by others has shown that fragmented signals have a peak frequency in the 25 to 50 Hz range.¹⁶ The 60 dB area chosen was well within the 72 dB range of the 12-bit A/D converter. The frequency scale was extended to 500 Hz to correspond to the range possible with the 512-point fast-Fourier transform. All data and FFTA calculations were performed with double-precision arithmetic (32 bits). Subsequently, FFTA data were transformed into single-precision arithmetic values (16 bits) and plotted. For patient-to-patient comparisons, the individual X, Y, and Z values for each region of interest was averaged. The 40 Hz intercept and the 60 dB area for the entire QRS complex, the terminal 40 msec of the QRS, ST segment, and T wave were expressed as means of the X, Y, and Z values.

Statistical analysis. Data are presented as mean ± SD. Group values are expressed as mean ± SEM. Statistical comparisons were performed with the unpaired t test. Intergroup and intragroup comparisons were performed by analysis of variance (ANOVA). Significance refers to a p value < .05.

Results

Validation. The time-domain representation of the ramp and 40 Hz sine burst functions used to simulate a small oscillatory waveform superimposed on the trailing edge of the QRS complex are shown in figure 1. To assure that no edge discontinuities existed, both functions were multiplied by a first-order Hanning window function before frequency-domain analysis. A comparison of results of the time- and frequency-domain analyses is shown in figure 2. The spectral analyses of the ramp function, 40 Hz sine burst, and the sum of the ramp and sine burst functions are shown along with both the original time-domain waveforms and the time-domain reconstruction from the Fourier series. As shown in figure 2, *C* the amplitude (0 to peak) of the sine burst was 0.1 mV and the peak amplitude of the ramp function was 1 mV. The time-domain perturbation that reflected the 40 Hz sine burst contribution was minimal but clearly detectable in the frequency domain. Moreover, the sine burst component was easily detected at levels as low as 10 μ V in the frequency domain as a bump in the spectra near 40 Hz. However,

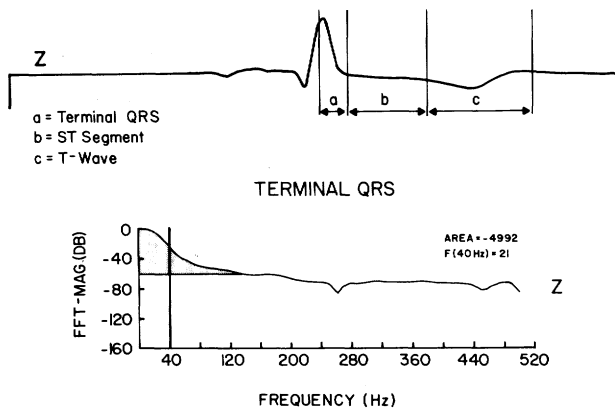


FIGURE 4. Representative example of a signal-averaged Z lead of the ECG with demarcation by the computer graphics cursor of the terminal 40 msec of the QRS complex, ST segment, and T wave. The FFTA of the terminal QRS is shown on the bottom, along with the computer-defined 40 Hz intercept (solid vertical line) and 60 dB area (shaded area).

it was not evident at these low levels in the time domain without prefiltering regardless of gain.

The effects of gain and prefiltering in the time domain are illustrated in figure 3. In each panel the

original time-domain waveform and idealized filtered waveform are shown. In the top panel the display gain was doubled when compared with figure 2, C. When the filter used in the reconstruction was 2 to 80 Hz, the filtered waveform and original waveform correlated closely. However, the sine burst contribution was still only minimally apparent. In the bottom panel the display gain was five times that in figure 2, C and the filter used in the reconstruction was 40 to 120 Hz. Thus, only after high-gain amplification and specific filtering was the 40 Hz sine burst adequately visualized in the time domain. A limitation of filtering is that the selection of the high- and low-pass filters requires a priori knowledge of the frequency content of the signal of interest. Furthermore, filtering may exclude potentially significant signals. Frequency-domain analysis avoids the inherent limitations of filtering used commonly for time-domain analysis.

Results of testing the portable signal-processing unit developed for use in patients showed an appropriate reduction of 20 dB in the system noise when 100 compared with one cycle were signal averaged. In addition,

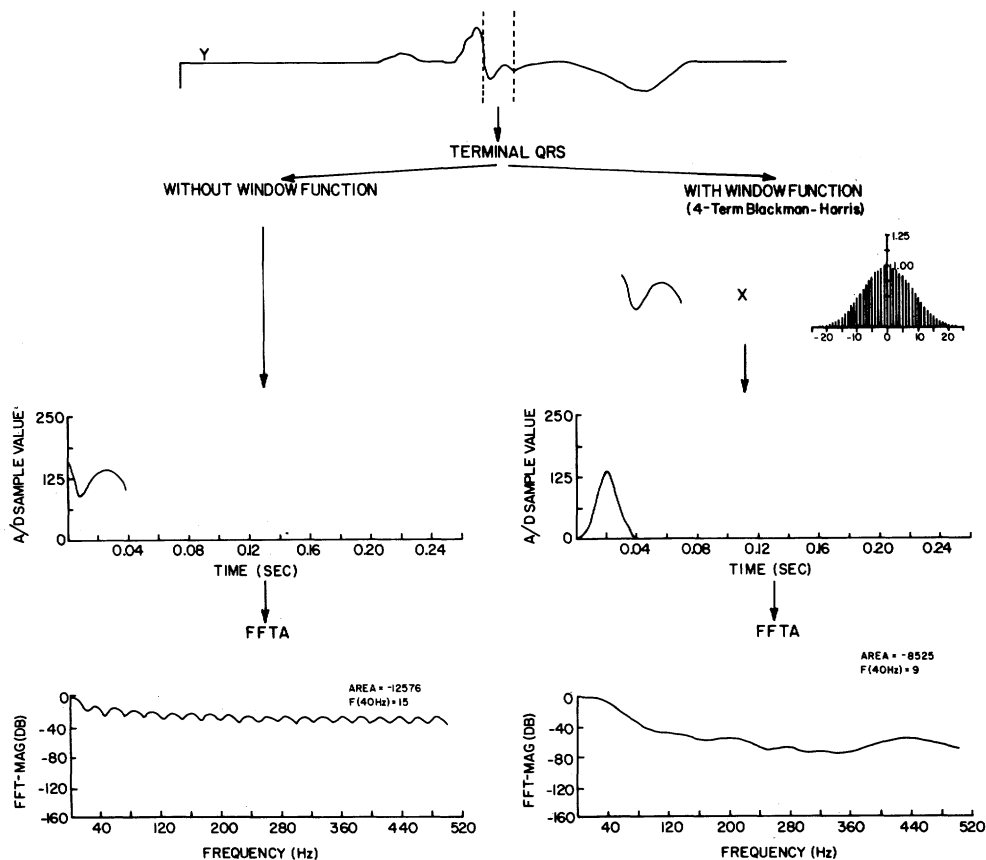


FIGURE 5. Parallel data processing of the terminal 40 msec of the QRS with (right) and without (left) the four-term Blackman-Harris window function. Left, The edge discontinuity between the initial and final sample point of the terminal QRS is readily apparent and adds artifactual frequency information to the signal of interest. As shown in the middle panel, right, multiplication by the four-term Blackman-Harris window eliminates these edge discontinuities and the initial and final sample points are now isopotential.

tion, a sine burst level as low as 2 μ V could be observed.

Studies in patients. FFTA of signal-averaged ECG leads X, Y, and Z showed significant ($p < .0001$) differences in the 60 dB area and the 40 Hz intercept of the terminal 40 msec of the QRS and ST segment in patients in group I compared with values in those in groups II and III. There were no significant differences in the 60 dB area or 40 Hz intercept of the terminal QRS or ST segment in patients with myocardial infarction without sustained VT when compared with normal subjects.

Representative plots of results of FFTA of the terminal QRS complex from a patient with prior myocardial infarction with sustained VT and from a patient with prior myocardial infarction without sustained VT are shown in figure 6. Each panel depicts power vs frequency plots of the terminal 40 msec of signal-averaged ECG complexes recorded from bipolar X, Y, and Z leads along with values for the 60 dB area and the decibel drop at the 40 Hz intercept. In each lead the terminal 40 msec of the QRS complex from the patient who had manifest sustained VT contained relatively more high-frequency components than the complex from the patient who had not, which was reflected by a greater value for the 60 dB area and a lesser decibel drop at the 40 Hz intercept. Figure 7 illustrates the FFTA of the ST segments from the same two patients. In each lead the ST segment in the signal from the

patient who had had an episode of sustained VT contained relatively more high-frequency components than the ST segment from the patient without VT. Spectral differences in the terminal QRS and ST segment were consistently most marked at frequencies lower than 120 Hz.

Since the current system configuration has a 12-bit A/D converter, a 72 dB drop is the maximal change that can be detected reliably. Comparisons between the mean 60 dB area and the 40 Hz intercept of the terminal 40 msec of the QRS complex and ST segment as well as those for the entire QRS complex and T wave for the three patient groups are summarized in table 2. There were no significant differences in the 40 Hz intercepts or 60 dB areas of the total QRS complex or of the T waves among the three groups.

Based on results of FFTA (mean \pm SD) of the terminal 40 msec of the QRS complex in normal subjects (group III), 60 dB area values greater than 2400 dB-Hz and 40 Hz intercept values less than 47 dB were defined as abnormal and indicative of an increase in the amplitude of high-frequency harmonics in the terminal QRS. Abnormal values for both the 60 dB area and 40 Hz intercept were found in 88% of patients with prior myocardial infarctions who had subsequent episodes of sustained VT (group I) and 15% of patients with prior myocardial infarctions without sustained VT (group II).

Based on results of FFTA (mean \pm SD) of the ST

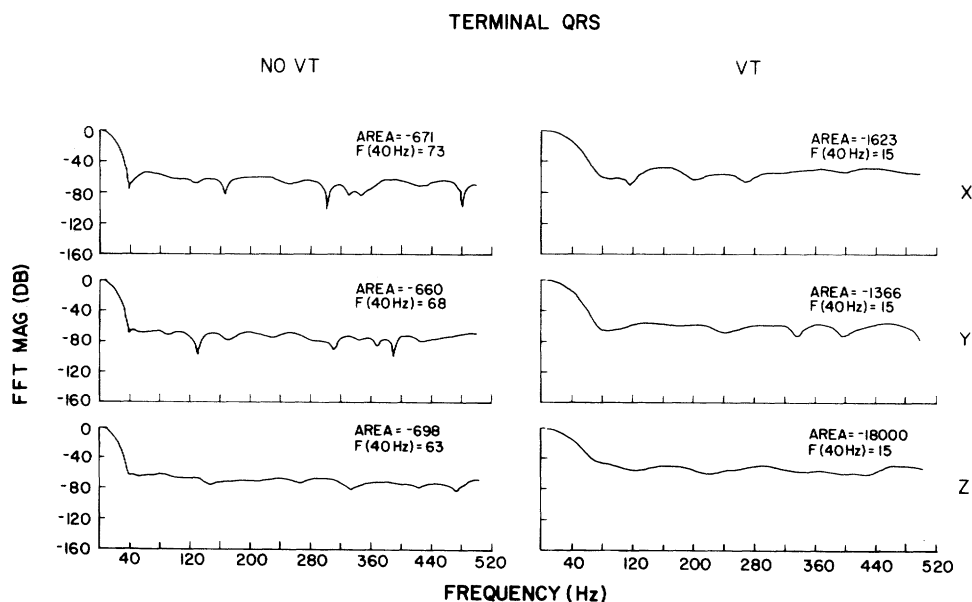


FIGURE 6. Representative FFTA of the terminal QRS complex from a patient with prior myocardial infarction without sustained VT (*left*) and from a patient with prior myocardial infarction and sustained VT (*right*). Shown in each panel are power vs frequency plots of the terminal 40 msec of signal-averaged ECG complexes recorded from bipolar X, Y, and Z leads, values for the 60 dB area, and values for the 40 Hz intercept. In each lead, the terminal 40 msec of the QRS complex from the patient with sustained VT contains relatively more high-frequency components than the complex from the patient without VT.

ST SEGMENT

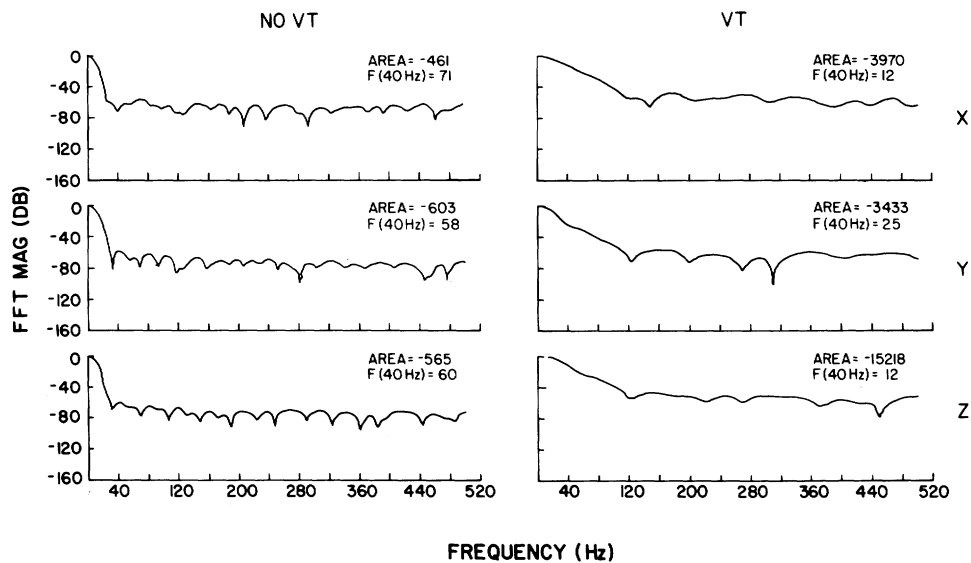


FIGURE 7. Representative FFTA of the ST segment from a patient with prior myocardial infarction without sustained VT (*left*) and from a patient with prior myocardial infarction and a subsequent episode of sustained VT (*right*). The display is similar to that in figure 4. In each lead, the ST segment from the patient with sustained VT contains relatively more high-frequency components than that from the patient without VT.

segment in group III subjects, 60 dB values greater than 2500 dB-Hz and 40 Hz intercept values less than 52 dB were defined as abnormal and indicative of increased amplitude of high-frequency harmonics in the ST segment. Abnormal values for the 60 dB area of the ST segment were found in 81% and 25% of group I and group II patients, respectively. Abnormal values for the 40 Hz intercept were found in 76% and 20%, respectively.

Values for both the 60 dB area and 40 Hz intercept of the terminal QRS and ST segment were independent of QRS duration, left ventricular ejection fraction, and complexity of spontaneous ventricular ectopy. Figure 8 illustrates the poor correlations between the values for the 40 Hz intercept of the terminal QRS and ST

segment and the QRS duration ($r = .2$) and left ventricular ejection fraction ($r = .19$) in group II patients. Similarly, values for the 60 dB area of the terminal QRS and ST segment did not correlate closely with QRS duration ($r = .25$) or left ventricular ejection fraction ($r = .16$). Analogous findings were obtained in patients in group I. In this group also, low correlation coefficients for the relationships between the 40 Hz intercept and QRS duration ($r = .07$) or left ventricular ejection fraction ($r = .36$) and those between the 60 dB area and QRS duration ($r = .12$) or left ventricular ejection fraction ($r = .13$) were observed. Figure 9 illustrates the poor correlation between the 40 Hz intercept and complexity of spontaneous ventricular ectopy in the 35 patients in group II. A similar lack

TABLE 2
Summary of results obtained with FFTA

	QRS	Terminal QRS	ST segment	T wave
60 dB area (dB-Hz)				
Group I	3721 ± 360	5300 ± 710 ^A	4350 ± 552 ^A	2023 ± 429
Group II	4057 ± 162	2207 ± 280	1749 ± 183	2682 ± 369
Group III	4569 ± 369	1791 ± 192	1661 ± 273	1213 ± 176
40 Hz intercept (dB)				
Group I	27 ± 1.6	28 ± 5.3 ^A	36 ± 5.6 ^A	55 ± 2.3
Group II	26 ± 0.7	46 ± 2.5	54 ± 2.2	53 ± 1.4
Group III	23 ± 5.2	53 ± 2.0	56 ± 1.2	61 ± 1.7

Data are expressed as mean ± SEM.

^Ap < .0001 compared with results in group II and group III patients.

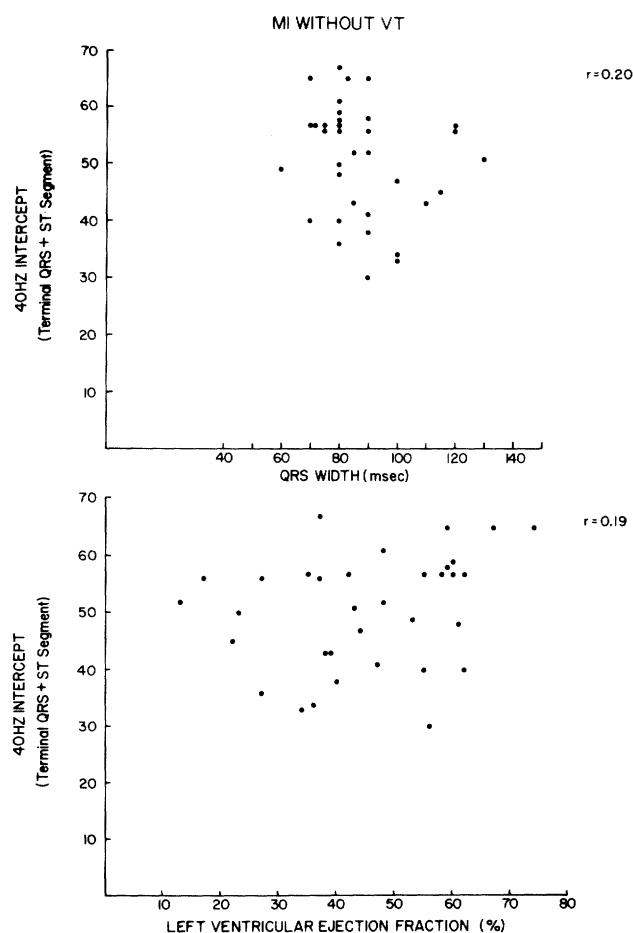


FIGURE 8. Poor correlation of the 40 Hz intercept with the QRS duration (*top*) and left ventricular ejection fraction (*bottom*) in patients with prior myocardial infarction without sustained VT or ventricular fibrillation. For convenience, the 40 Hz intercept values for the terminal QRS and ST segment were averaged.

of correlation was observed between the 60 dB area and the complexity of spontaneous ventricular ectopy.

Discussion

Conventional surface ECGs during arrhythmia-free intervals do not identify patients prone to manifest sustained VT despite the occult delayed activation presumably present. Recently, several groups have used signal-processing techniques in the time domain to identify low-amplitude potentials in the terminal portion of the QRS complex and ST segment of signal-averaged ECGs from patients and experimental animals with sustained VT.¹⁻⁸ Although the high-frequency, low-amplitude potentials recorded from the surface ECG may reflect either delayed or disorganized ventricular activation,⁹⁻¹¹ the analytic methods used generally have been dependent on high-gain amplification and complex filtering and therefore unavoidably prone to some distortion and subjectivity.

To circumvent some of these limitations, we devel-

oped, tested, and implemented a portable signal-processing system using signal-averaging techniques and FFTA. FFTA is a powerful computer-based analytic method for signal processing in the frequency domain and it allows some of the inherent limitations of time-domain analysis to be avoided, including the a priori selection of filters with its risk of excluding signals of potential interest.¹² The results obtained with both a computer-generated mathematical model and an analog test voltage demonstrated that FFTA facilitates objective identification and characterization of low-amplitude, oscillatory potentials in signal-averaged ECGs. Results obtained in patients with prior myocardial infarctions and subsequent episodes of sustained VT demonstrated significant ($p < .0001$) differences in the frequency content of the terminal QRS complex and ST segment of sinus beats when compared with results in patients with prior myocardial infarctions without sustained malignant arrhythmias and with those in normal subjects.

Several major sources of artifactual frequency shifts must be excluded before reliable differences in spectral content in signals from patients with and without sustained VT can be accepted. The first and most obvious is the contribution of random background noise, which in our current system configuration is reduced to 1.5 μV by signal averaging. A second and more subtle source of possible artifact relates to introducing step discontinuities between regions of interest (terminal

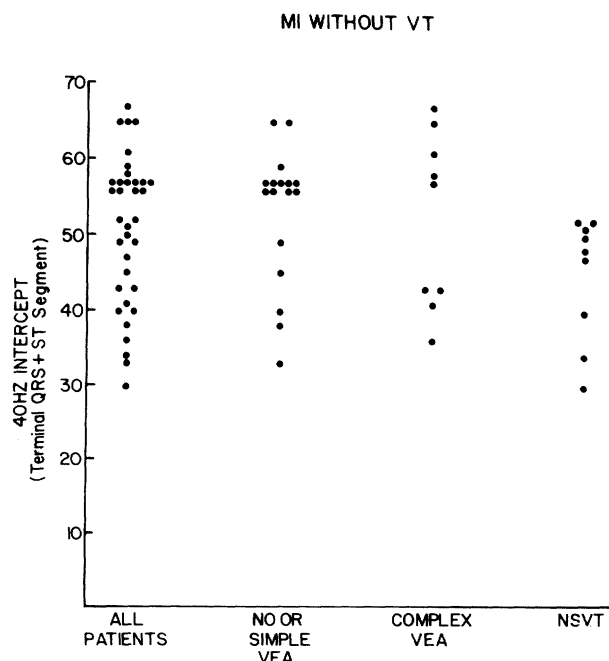


FIGURE 9. Poor correlation of the 40 Hz intercept with the complexity of spontaneous ventricular ectopy in patients with prior myocardial infarction without sustained VT or ventricular fibrillation.

QRS, ST segment, and T wave). The use of Fourier analysis requires the assumption that the signal contained in a sample window interval is a periodic function. However, if the initial and final sampling points are not at zero potential, a sharp discontinuity will be introduced between the end of one cycle and the beginning of the next that will artifactually add both high- and low-frequency harmonics to the original signal. To obviate this source of error, mathematical window functions are used that smooth the windowed data to zero at the boundaries. Since multiplication by the window function in the time domain leads to convolution in the frequency domain, care must be taken to choose a window function that allows detection of nearby components of significantly different amplitudes without compromising resolution, dynamic range, or ease of implementation. Accordingly, a four-term Blackman-Harris window function was used in our system.¹³

In this study, FFTA was performed on the entire QRS complex, terminal 40 msec of the QRS complex, ST segment, and T wave of signal-averaged X, Y, and Z ECG leads. The 60 dB area and 40 Hz intercept of each region of interest were compared in signals from patients with prior myocardial infarction and subsequent spontaneous sustained VT, patients with prior myocardial infarction and spontaneous ventricular ectopy but no sustained VT, and normal subjects. Results of FFTA demonstrated significant differences ($p < .0001$) for both the 60 dB area and 40 Hz intercept of the terminal QRS and ST segment in signals from patients with prior myocardial infarction and subsequent sustained VT compared with those from patients with prior myocardial infarction without sustained VT and from normal subjects. Increases in the amplitudes of high-frequency components in the terminal QRS were found in 88% of patients with sustained VT in contrast to in 15% of patients with prior myocardial infarction without sustained VT.

Recent studies with time-domain analysis have used a variety of high- (25 to 100 Hz) and low- (250 to 300 Hz) pass filters.¹⁻⁸ A potential limitation of complex filtering is the lack of a priori knowledge of the frequency distribution of the signals of interest and thus the inherent risk that use of filtering during data analysis will exclude signals of potential interest. Comparisons of results of FFTA in patients with and without sustained VT indicated that the major difference in frequency content of the terminal QRS and ST segment was at frequencies less than 120 Hz. Judging from these results, it appears that previous studies in which time-domain analysis and prefiltering were used may

have excluded some signals of potential interest. Varying use of prefiltering may explain some of the divergent results from other studies seeking to detect low-amplitude, high-frequency components in the terminal QRS and ST segment.

The bandwidth of our analog amplifiers extends from 0.05 to 470 Hz. However, some inherent limitations of frequency-domain analysis and signal averaging can alter the ability to definitively detect discrete signals over this broad range of frequencies. First, frequency resolution is inversely proportional to sample size (entire QRS complex, terminal 40 msec of the QRS, ST segment, and T wave). Frequency resolution is 25 Hz for the terminal 40 msec of the QRS complex and 6.6 Hz for a 150 msec ST segment. We analyzed each of these components individually to compare our results with those from studies in which time-domain techniques were used.¹⁻⁸ Frequency resolution could be enhanced by examining larger sample sizes (i.e., terminal 40 msec of the QRS plus the ST segment) or by increasing the sampling rate. Nevertheless, our initial approach proved to be capable of providing reliable assessment of the range of frequency patterns for each region of interest.

A second inherent limitation is attenuation of high-frequency signals during signal averaging due to trigger jitter of biological signals, a limitation encountered with time and frequency-domain analyses. Trigger jitter is ± 1 msec in our system and comparable to the 0.5 to 2.0 msec jitter reported by other investigators.¹⁻⁸ Jitter in this range can operate as a low-pass filter and mask high-frequency signals. In our system, with the worst case of reference jitter being ± 1 msec, the -3 dB point is 134 Hz.¹⁷ However, as mentioned previously, the major differences in the terminal QRS and ST segment were observed at frequencies less than 120 Hz.

In the present study, significant differences in the frequency content of both the terminal QRS and ST segment were observed consistently in patients with and without sustained VT. The electrophysiologic alterations responsible for the abnormal signals detected by frequency-domain analysis have not yet been elucidated. Abnormalities detected by time-domain analysis appear to represent delayed or disorganized ventricular activation.⁹⁻¹¹ High-pass filters have been used to reject low frequencies in the ECG associated with the plateau and repolarization phases of the action potential, ST segment, and T wave. The contribution of these phases to the spectral differences observed in the present study in patients with and without VT have not yet been delineated.

Results of FFTA were independent of QRS duration, left ventricular ejection fraction, and the complexity of spontaneous ventricular ectopy. Patients with prior myocardial infarction without spontaneous sustained VT manifested simple as well as complex ventricular ectopy, including nonsustained VT. Most studies in which time-domain analyses were used have been restricted to comparisons between patients with sustained VT and patients with absent or simple ventricular ectopy,⁵ normal subjects,⁸ or patients in whom the frequency and complexity of ventricular ectopy was not specifically characterized.⁷

Judging from the results of this study, FFTA facilitates delineation of concealed information in signal-averaged ECGs. It appears particularly well suited for definitive characterization of the frequency content of late potentials, which to date have been detectable only with signal-processing techniques in the time domain. Accordingly, FFTA offers particular promise for accurate noninvasive detection of patients prone to sustained VT while allowing some of the inherent limitations of high-gain amplification and a priori filtering to be avoided.

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